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# PROVISIONAL APPLICATION FOR PATENT COVER SHEET

This is a request for filing a PROVISIONAL APPLICATION FOR PATENT under 37 CFR 1.53(c).

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**TITLE OF THE INVENTION (500 characters max)**

**MULTI-SEGMENT CONE-BEAM RECONSTRUCTION ALGORITHM FOR TOMOSYNTHESIS IMAGING**

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[Page 1 of 1]

*Respectfully submitted,*

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***USE ONLY FOR FILING A PROVISIONAL APPLICATION FOR PATENT***

PROVISIONAL APPLICATION

FOR

UNITED STATES PATENT

TITLE OF INVENTION

# MULTI-SEGMENT CONE-BEAM RECONSTRUCTION ALGORITHM FOR TOMOSYNTHESIS IMAGING

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# MULTI-SEGMENT CONE-BEAM RECONSTRUCTION ALGORITHM FOR TOMOSYNTHESIS IMAGING

## FIELD OF THE INVENTION

The present invention relates to a system and method for imaging a target element using tomosynthesis. More specifically, the invention relates to a system, method and computer program product for creating a three-dimensional image of target elements from a plurality of radiation absorbance projection images taken from different angles.

## BACKGROUND OF THE INVENTION

Imaging of a patient's tissue has become a common screening and/or diagnostic tool in modern medicine. One example of such imaging is mammography, or the imaging of a patient's breast tissue. Breast cancer remains the most common cancer among women today, however, at this time there is no certain way to prevent breast cancer and the best strategy for dealing with breast cancer is early detection of the cancer so that it may be treated prior to metastatic spread. Accordingly, it is important for patients to have access to imaging techniques and systems that will detect very small cancers as early in their development as possible.

A three-dimensional imaging approach called "tomosynthesis" has been developed (see United States Patent No. 5,872,828, which is incorporated herein by reference for its teachings relating to tomosynthesis systems and methods) which shows great promise for early detection of cancer. Tomosynthesis allows the reconstruction of topographic planes on the basis of the information contained in a series of projections acquired from a series of viewpoints about the target object. They need not be regularly spaced, numerous, or arranged in any regular geometry. The tomosynthesis technique has been demonstrated to provide improved spatial differentiation of overlapping and nearby tissues at very high resolution comparable to projection 2D imaging, with approximately comparable radiation dose.

The problem of 3D reconstruction from tomosynthesis projections is difficult one. One promising technique for 3D reconstruction from tomosynthesis projections is provided in United

States Provisional Patent Application Serial No. 60/446,784, which is incorporated herein by reference. This technique applies a cone-beam geometry in an iterative forward-projection and back-projection method based on maximum-likelihood estimation of volumetric distribution of attenuation coefficients, using an estimation-maximization algorithm. However, the amount of computing power required to perform the 3D reconstruction will, for typical inexpensive computer systems, result in significant delays before the 3D reconstruction is available.

## DETAILED DESCRIPTION OF THE INVENTION

The systems and methods of the present invention improve upon systems and methods known in the art by providing tomosynthesis apparatus and techniques for three-dimensional imaging of target elements that overcome the problems of conventional three-dimensional imaging systems, including known tomosynthesis systems. The present invention enables the use of tomosynthesis to efficiently provide accurate three-dimensional imaging of a target element in a shorter time than has previously been possible. The algorithm is efficiently tuned by computationally appropriate segmentation of the target volume and directs the reconstruction of volume segments to a multiplicity of CPUs within a multi-CPU computer cluster so that the volume segments can be reconstructed in parallel. In a preferred embodiment, the volume segments are selected to be optimally overlapping in order to provide mutual boundary coverage of what would otherwise be under-determined edge volume segments. Thus it increases the speed of the overall volume reconstruction while providing a quality of reconstruction that is substantially the same as for reconstructions that do not employ the segmentation algorithm of the invention. The invention is applied below to one preferred embodiment in which the system is used for tomosynthesis mammography; however, the invention will be useful in a variety of three-dimensional imaging situations. For example, the invention can be applied to a variety of patient imaging problems such as heart imaging, or imaging of the soft tissues or bones of the hand. The imaging system of the invention can be used for diagnoses (as is described below for tomosynthesis mammography) or it may be used for other applications such as three-dimensional modeling for the purpose of fitting an implant (whether orthopedic, such as a hip or knee implant, an artificial heart, or other type of implant) or for use in surgical navigation systems.

What follows is a description of one preferred embodiment of the invention:

1. Introduction to an Exemplary Tomosynthesis Mammography System

Tomosynthesis mammography is a three-dimensional breast imaging technique. It involves acquiring projection images of a breast at a plurality of viewpoints, typically over an arc or linear path. Three-dimensional distribution of x-ray attenuation coefficient of the breast volume is reconstructed from these projections. A prototype Tomosynthesis system for breast imaging has been developed by Massachusetts General Hospital (MGH) and General Electric Cooperate Research and Development (GECRD). Eleven projections are acquired by moving the x-ray tube over a 50° arc (-25° to +25°) above the breast in 5° angular steps. Breast and detector are stationary during the image acquisition **(Figure 1A)**.

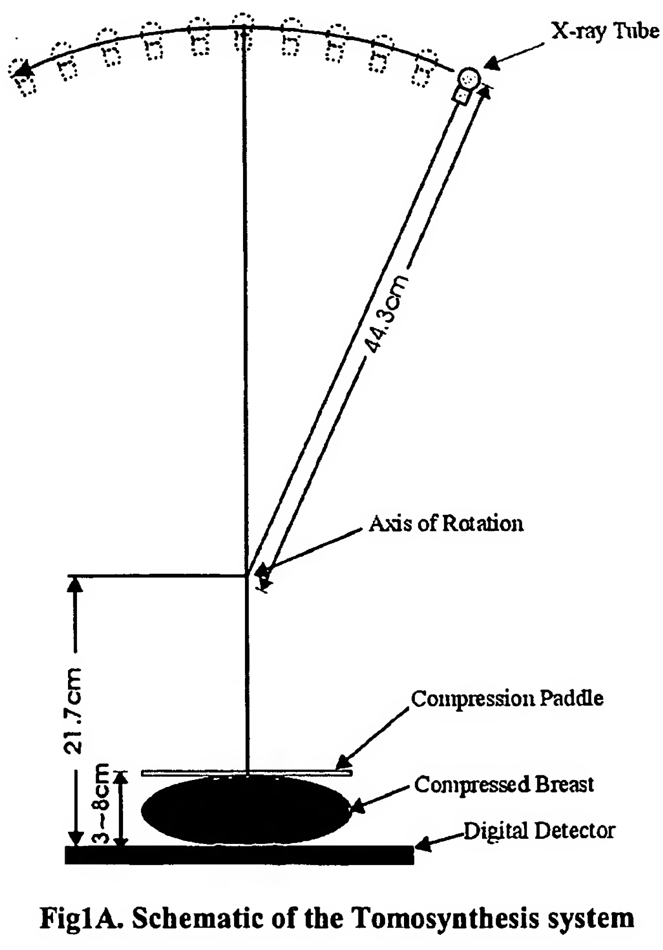



**Fig1A. Schematic of the Tomosynthesis system**

Certain characteristics of this exemplary embodiment of a tomosynthesis system useful with the invention are described below:

Spatial resolution and contrast resolution: The Tomosynthesis system can use an amorphous-Silicon-based flat panel detector on which a CsI crystal phosphor is grown epitaxially read out as 2304X1800 pixels (100μm pixel pitch) via a TFT array. This particular detector has a linear response over exposure levels up to 4000mR and 12 bits of working dynamic range. Each plane of the 3D reconstruction has about the same resolution as the detector (100um) but the depth resolution is on the order of a millimeter.

Dose: The target/filter combination is Rh/Rh and the accelerating potential is 25~33kVp to image breasts with 3~8cm range of thickness. The total x-ray dose for acquiring 11 projections is approximately 1.5 times of that used for one film-screen mammogram. Each projection is a low dose breast image (approximately 1/11 of the does per projection).

Patient motion: Patient motion is reduced by fast image acquisition. Using cone-beam x-ray geometry and area detector, a projection of the whole breast can be recorded with one x-ray exposure at each angle. For each projection, the exposure time is 0.1~0.2s and detector readout time is about 0.3s. Rotation to the next angle is performed during the detector readout. The total image acquisition time for 11 projections is about 7sec. Breast compression also helps to reduce patient motion.

Image acquisition geometry: The design of the Tomosynthesis system is based on the conventional mammography system. The MLO views have been used in most cases since it provides the most complete coverage of the whole breast.

Digital tomosynthesis mammography is particularly well suited for removing the tissue-overlap, and therefore reducing false-positive and false-negative diagnoses of breast cancer. The geometry of the tomosynthesis mammography system is further illustrated in **Figure 1B** which illustrates two orthogonal views of the tomosynthesis system geometry with the left hand view

being along the patient's chest wall (with the X-ray source traveling into and out of the page) and the right hand view being in a direction toward the patient's chest wall. The system acquires 11 projection images of the breast and then reconstructs an estimate of the volumetric distribution of attenuation coefficients likely to have resulted in the measured projections.. Each projection image consists of 2304x1800 100 μm pixels. The thickness of compressed breasts ranges from <25 mm to >80mm, typically within this range. The chest-to-nipple distance ranges from <50mm to >160mm,but typically lies within this range. The reconstructed volume distribution consists can be represented by slices spaced 1 mm apart with 100x 100 μm in-plane voxels. Therefore, the size of reconstruction image is (30~80) x 2304 x (500~1600). Using a PC with a 2.4GHz CPU, the reconstruction of a breast volume takes up to 5 hours. This is far to long for clinical use. The goal of this invention is to reduce the reconstruction time to a few minutes. This will not only permit clinical care, but will allow "real-time" needle placement to guide biopsies, etc.

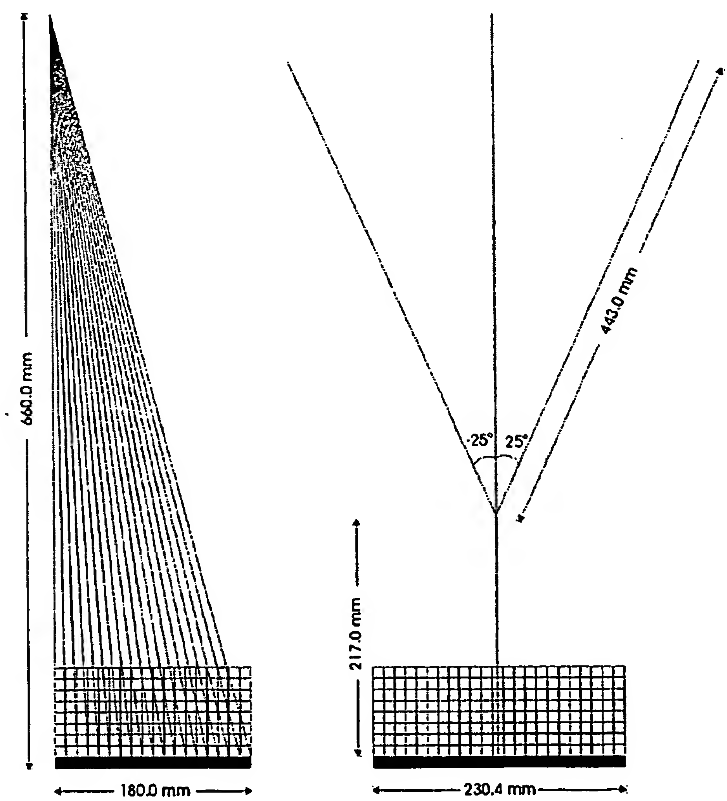



**Figure 1B. Tomosynthesis Mammography System Geometry**

## 2. Technical Description of the Invention

A parallel computing reconstruction can be developed and implemented on a computer cluster with, for example, 32-64 processors. In parallel computing reconstruction, the computation task is divided into independent smaller tasks, each of which reconstructs a part of the whole breast accomplished by one processor of the computer cluster. The volume image of the whole breast can be retrieved from the results of the small tasks.

Projection images are divided into a series of stripe-like segments, parallel or curvilinear oriented along the chest wall, from the chest wall to the nipple. Each projection segment covers part of the whole breast volume. With appropriate segmentation, projection segments will continuously inform the whole breast volume. The complete or partial volume can be composed by careful assembly and blending results of segment reconstructions. Increasing the number of segments results in smaller segmented volume coverage and faster reconstruction.

### Geometry of Non-Overlapping Segmentation

One method of segmentation is shown in **Figure 2**, provides a virtual view that is from the same perspective as the left hand diagram in **Figure 1B**. The vertical axis is the breast thickness (Z = <3->8cm, but generally between) dimension and the horizontal axis is the chest-to-nipple direction (column 100-1700 on a specific detector, Y = 1-17cm). The motion of the X-ray tube is generally perpendicular to this plane, with intersection at 0° (Z=66cm at 0° and Z=61cm at ±25°). In this example, the projection image is divided into continuous, non-overlapping segments (the segments being represented by the columns graphically illustrated in **Figure 2**) where each segment contains 100 columns. The volume coverage by each projection segment is shown by connecting the two edges of a segment (the bottom portion of each segment column illustrated in **Figure 2**) to the X-ray source (the right-most line of each pair of lines (the pink line in the original graphic) connects to the source at 0° and the left-most line of each pair of lines (the green line in the original graphic) connects to the source at 25°). There is a mismatch between the volume coverage by 0° source and that by 25°. This mismatch is bigger for thicker breasts and for segments further away from the chest wall (**Table 1**). In the worst case (8cm thick breast, 16cm from the chest wall), this mismatch is ~16 pixels. The mismatch of volume coverage indicates that, with this non-overlapping segmentation method, a small portion of the reconstructed volume

(at the edge of the volume) will suffer from "missing projection data" within the segmented projection set.

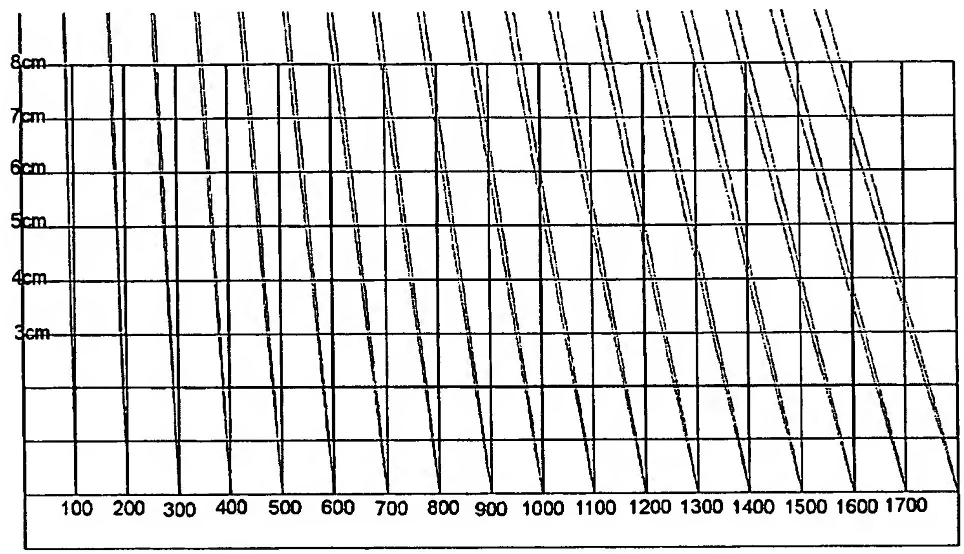


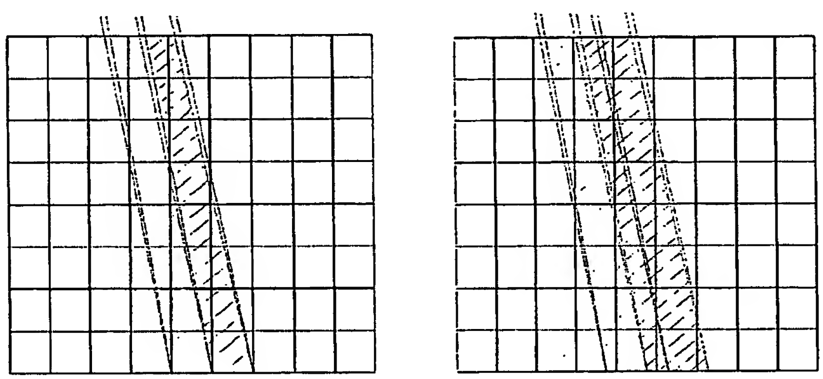

**Figure 2. Segmented Projections and Volume Coverage**

| Column | 3cm | 4cm | 5cm | 6cm | 7cm | 8cm |
|---|---|---|---|---|---|---|
| 0 | 0 | 0 | 0 | 0 | 0 | 0 |
| 100 | 0 | 0 | 1 | 1 | 1 | 1 |
| 200 | 1 | 1 | 1 | 1 | 2 | 2 |
| 300 | 1 | 1 | 2 | 2 | 3 | 3 |
| 400 | 1 | 2 | 2 | 3 | 3 | 4 |
| 500 | 2 | 2 | 3 | 4 | 4 | 5 |
| 600 | 2 | 3 | 4 | 4 | 5 | 6 |
| 700 | 3 | 3 | 4 | 5 | 6 | 7 |
| 800 | 3 | 4 | 5 | 6 | 7 | 8 |
| 900 | 3 | 4 | 6 | 7 | 8 | 9 |
| 1000 | 4 | 5 | 6 | 7 | 9 | 10 |
| 1100 | 4 | 5 | 7 | 8 | 10 | 11 |
| 1200 | 4 | 6 | 7 | 9 | 10 | 12 |
| 1300 | 5 | 6 | 8 | 10 | 11 | 13 |
| 1400 | 5 | 7 | 9 | 10 | 12 | 14 |
| 1500 | 6 | 7 | 9 | 11 | 13 | 15 |
| 1600 | 6 | 8 | 10 | 12 | 14 | 16 |

**Table 1. Mismatch of Volume Coverage by 0° and 25° Sources (with Pixel as Unit)**

<u>Geometry of Overlapping Segmentation</u>

In the example of non-overlapping segmentation, a small region of "bad volumes" occurs at the edge of the reconstructed volume. In **Figure 3A**, the areas marked by cross-hatching (yellow and green lines in the original graphic) represent "good volumes" and the gap between them represents the "bad volume". Overlapping segmentation can be used to solve this problem. In this method, a projection segment overlaps with its neighbor segments so that the volume coverage also overlaps (**Figure 3B**, areas marked by cross-hatching (yellow and green lines in the original graphic)). If the overlap of projection segments is big enough, "bad volumes" will be located only in the overlapping regions (**Figure 3B**). Therefore, a "bad volume" region in one segment may overlap with a "good volume" region in a neighbor segment. Data describing the entire reconstructed volume can then be retrieved by using only "good volumes" and ignoring redundant "bad volumes".



**(A) Non-Overlapping Segmentation** **(B) Overlapping Segmentation**

**Figure 3. Non-Overlapping and Overlapping Segmentation**

## 3. Preliminary Results: 50-Row Segments and 50% Overlapping

### Projection Segmentation

An example was implemented using Segmented reconstruction with 50-row segments and a high 50% overlapping. Projections were divided into multiple segments from the chest wall edge to the nipple edge (**Figure 4**, with the chest wall at the bottom of the figure, the geometry thus being rotated 90 degrees counter-clockwise with respect to **Figures 1B** (left hand side), **2**, **3A** and **3B**). Each segment consisted of 50 detector rows. The first 25 rows of segment N overlapped with the last 25 rows of segment N-1 and the last 25 rows of segment N overlapped with the first

25 rows of segment N+1. The total number of segment was determined by the size of the breast from chest wall to nipple. If the projection consists of M rows, the number of segments will be 2×(M/50)-1. All the eleven projections were segmented in the same way.

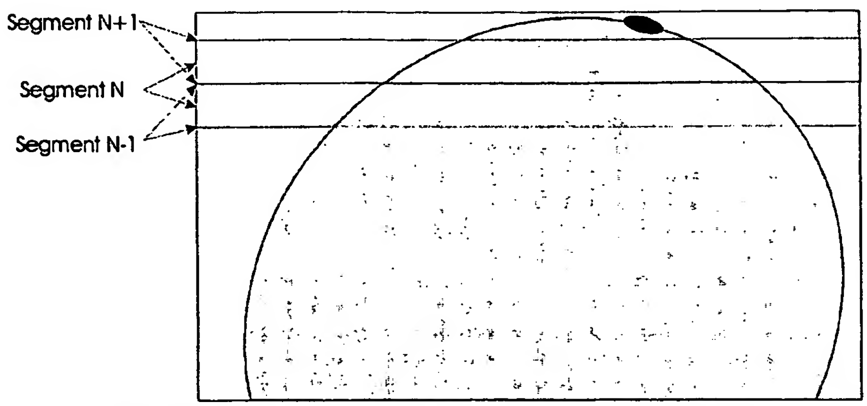



**Figure 4. Segmentation of Tomosynthesis Projection**

The eleven projection segments with the same distance to the chest wall side were grouped into a set and used to reconstruct a volume segment. Reconstructed volume segments from all projection segments were then merged to form the whole breast volume. The reconstruction volume segment from a projection segment had a "slanted rectangular" shape (**Figure 5**). The slope of the "slanted rectangular volume" was determined by the location of projection segments used to reconstruct this volume (**Figure 5**). A reconstructed volume segment had "bad voxels" close to its boundaries because the mismatch of volume coverage by between projection segments at different angles (**Table 1**). However, the consecutive projection segments were overlapped by 50%, so the boundary of one segment was close to the center of a neighbor segment. The corresponding reconstructed volumes overlapped in a similar way and this 50% overlap was sufficient to make all "bad voxels" in overlapping regions. Therefore, a location taken by a "bad voxel" in one volume segment was overlapped by a "good voxel" in a neighbor volume segment so the whole breast volume can be retrieved with "good voxels".

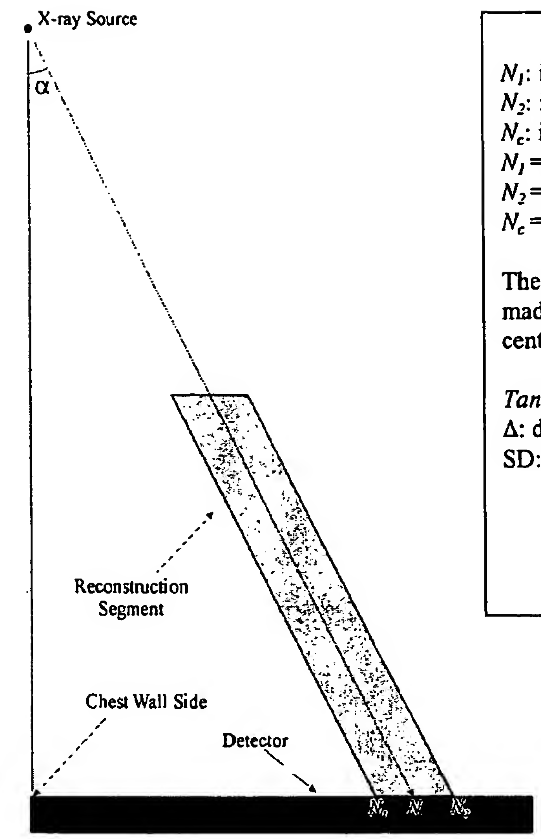



Figure 5. Reconstructed Volume Segment

$N_1$: index of the first detector row in segment $N$
$N_2$: index of the last detector row in segment $N$
$N_c$: index of the center row in segment $N$
$N_1 = 25 \cdot N$
$N_2 = 25 \cdot N + 49$
$N_c = 25 \cdot N + 25$

The slope of the volume is represented by α, the angle made by the detector plane and the line connecting the center of segment $N$ and the X-ray source at 0°:

$Tan(\alpha) = (Nc \cdot \Delta)/SD$
Δ: detector pixel size
SD: source-to-detector distance

**Figure 5. Reconstruction Segmentation**

Reconstruction of Segmented Volume

A segment of reconstruction volume was stored in a 3-D image array (the dotted volume in **Figure 6**), consisting of parallel image slices (**Figure 6**). However, there was a shift between slice centers because the volume had a "slanted rectangular" shape (the line connecting the centers of image slices points to the X-ray source). The implementation of the reconstruction algorithm was almost the same as that for the conventional reconstruction method, except that an extra operation (shift of slice center) was now taken in the calculation from volume index (i, j, k) to coordinate position (x, y, z) and vice versa. The value of the shift was determined by the slice position (the distance above the detector and the slope of the slanted volume) and the slope of the slanted volume varied for each volume segment (represented angle alpha in **Figure 5**).

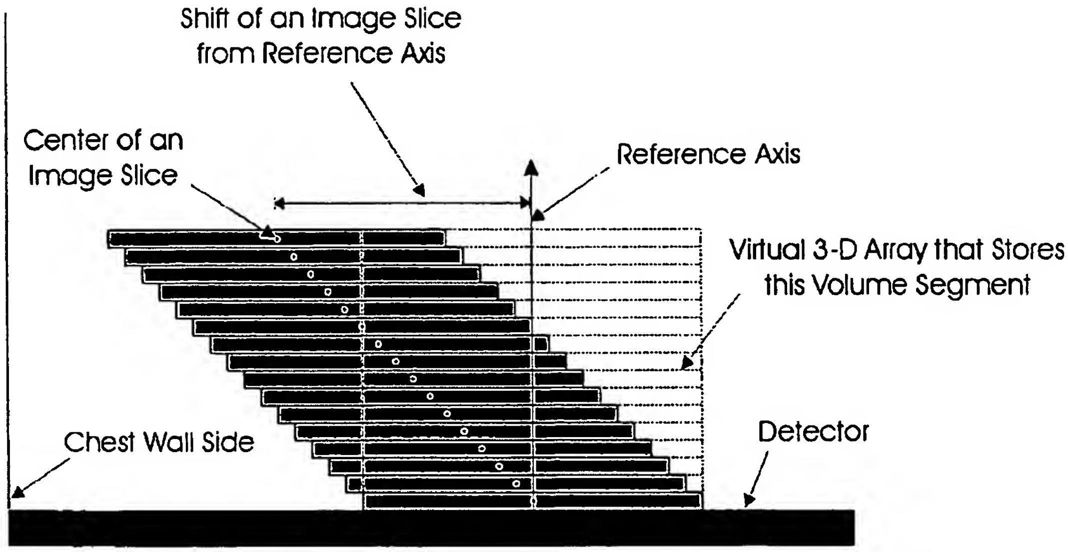



**Figure 6. Reconstruction of a Volume Segmentation**

Merging of Reconstructed Volume Segments

When all the volume segments were reconstructed, they were merged to form the image of the whole breast volume. **Figure 7** shows the merging of two neighbor volume segments. The two reconstructed segments had about 50% overlapping similar to the overlapping in projection segments. Bad voxels were in regions close to volume boundaries. However, the "bad voxel region" in one segment was covered by the "good voxel region" in a neighbor segment due to overlapping. The whole breast volume was then retrieved using only "good voxel regions".

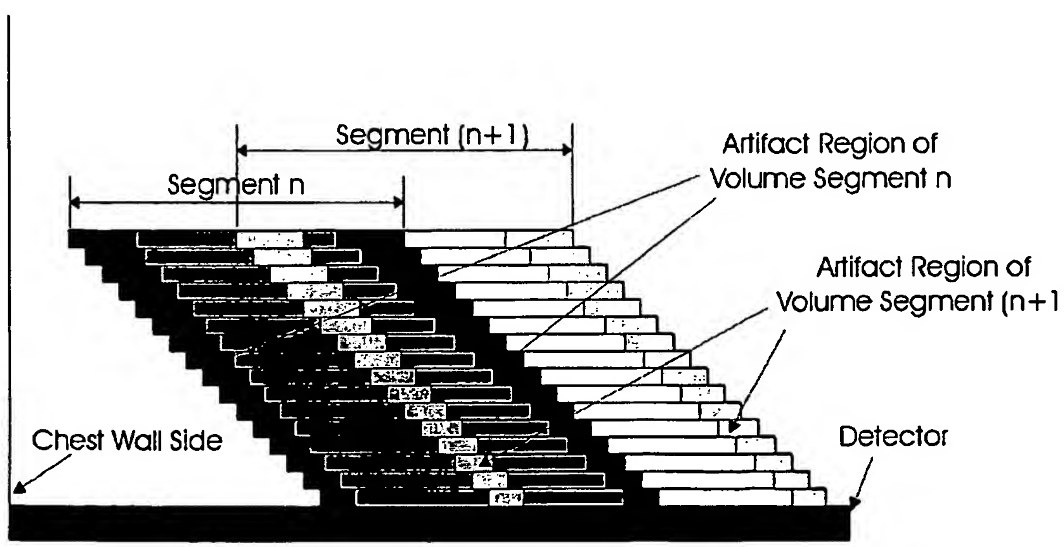


Figure 7. Mergence of Volume Segmentations

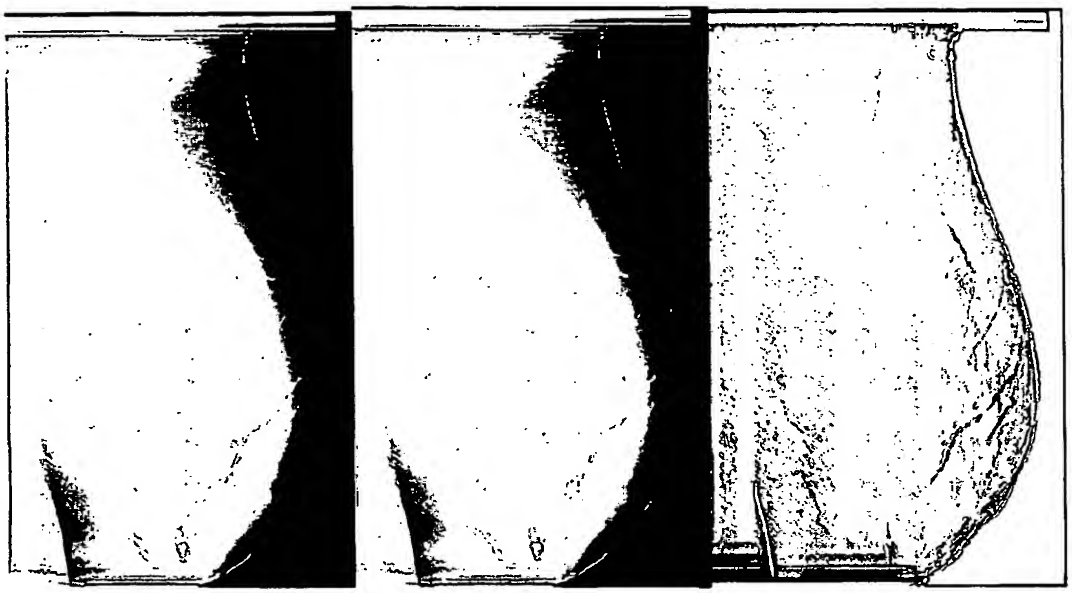
3. Performance of the Invention

A Tomosynthesis patient case (case number: 5131-Right) was selected for performance test of the segmented reconstruction. The thickness of the breast was 7cm and the distance from the chest wall to nipple was ~14.5cm. This was one of the largest breasts among those that had been imaged. Reconstructions were done with both a conventional (unsegmented) algorithm and the segmented algorithm for 9 iterations. The reconstruction volume was 2304x1450x70 (height, width and thickness) with a total size of ~450MB. The workstation for reconstruction had a PIII 866MHz processor and 4GB RAM.

For the conventional algorithm, it took ~53 min/iteration and totally 8 hours for 9 iterations. For the segmented reconstruction algorithm, each projection was divided into 57 (2x[1450/50]-1) segments with a 50-row width. The first 25 rows of a segment was overlapped with the second 25 rows of the neighbor segment before it; and the second 25 rows of this segment was overlapped with the first 25 rows of the neighbor segment after it. The projection segments from different projections were grouped according to the distance from the chest wall. Therefore, 57 sets of projection segments were generated, each containing 11 projection segments with the same distance to the chest wall. Each set of the projection segments was used to reconstruct a volume segment. Each volume segment had 2304x50x70 voxels and took ~19 minutes for the total nine iterations (~2.1min/iteration). Currently, the mergence of reconstruction segments takes ~5 minutes for this case. These reconstructions were done one by one using a single computer processor so the total reconstruction time is the summation of time used for all segments. When implementing the algorithm of the invention on a computer cluster, the total reconstruction time will be mainly determined by the time for the reconstruction of one segment.

Reconstructed volumes were compared. Four sets of reconstructed volumes were presented as slices (Z=10, 30, 50 and 70mm) and are shown below (**Figure 8**). The images in the left column are from the conventional reconstruction algorithm, the images in the middle column are from the segmented reconstruction algorithm and the images in the right column are the difference between the conventional reconstruction and the segmented reconstruction. The value

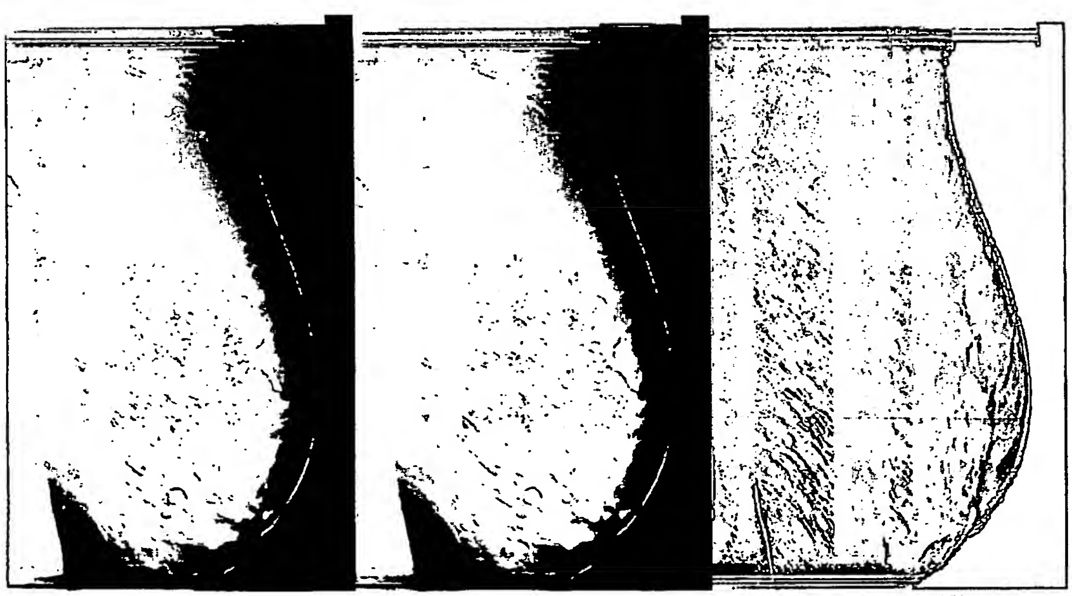
of most pixels of the difference images is less than 1% of the value of the corresponding reconstruction image pixel. The result demonstrates that images from the segmented reconstruction algorithm provide equivalent diagnostic information as those from the conventional reconstruction algorithm.



(A) (B) (C)

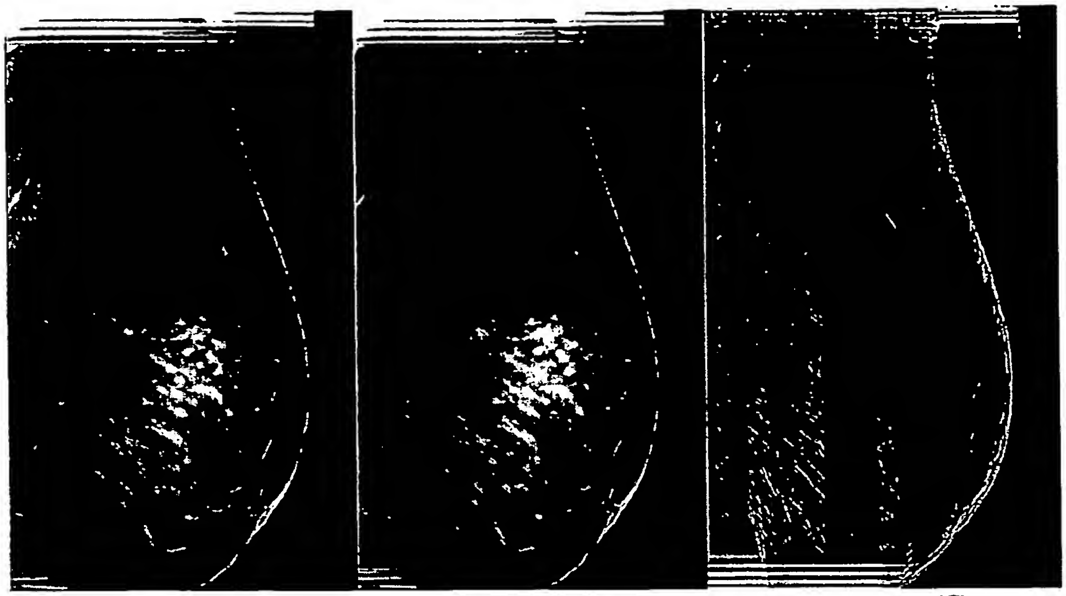
Figure 8.1 Image of Segmented Reconstruction: Z = 10mm





(A) (B) (C)

Figure 8.2 Image of Segmented Reconstruction: Z = 30mm



(A) (B) (C)

Figure 8.3 Image of Segmented Reconstruction: Z = 50mm

(A) (B) (C)

Figure 8.4 Image of Segmented Reconstruction: Z = 70mm



4. Conclusion

The segmented reconstruction method on a single processor is working successfully. In the test with a large-size breast, it took 2.1 min/iteration for reconstruction with a 50-row projection segment and it took ~5 minutes to merge the reconstruction segments to form the whole breast volume. The image quality of volumes formed from the segmented reconstruction

method is equivalent to that from the conventional reconstruction method. If this is implemented on a computer cluster with the same CPU (Pentium III, 866MHz, 133MHz bus speed, 133MHz RAM speed) and there is a sufficient number of CPUs, the time for the whole reconstruction will be the same as the time for reconstructing one segment due to parallel processing. The total time for this case will be

$$T = (2.1 \text{ min/iteration}) \times (\text{iterations}) + 5\text{min merging time} + \text{data transfer time}$$

The first two terms take ~25 minutes for this case on the current computer. Compared with 8 hours conventional reconstruction, segmented reconstruction reduces the time by a factor of 20. The typical breast size of a patient is about the half of this testing case.

Currently, computer clusters with 3 times faster processors (above 2.2GHz, 266MHz RAM, 533MHz bus speed) are available. With such a system, we expect the total time (reconstruction + merging + transferring) for a case to be reduced to ~5 minutes for normal breast size and <10 minutes for very large size.

Advantages of the invention thus include fast computation speed for clinical applications with no loss of image quality. In addition, these performance increases can be provided with no increased cost at installations that already include multi-CPU computer clusters. Additionally, specialty FPGA arrays can be fabricated to implement multiple CPUs on a single PC add-in card. One or a series of these cards may employ the same strategy outlined above.

A person of ordinary skill in the art will appreciate further features and advantages of the invention based on the above-described embodiments. For example, specific features from the references incorporated by reference above may be incorporated into systems, methods or computer program products of the invention as well as features referred to in the claims below which may be implemented by means described herein and described in those documents. Accordingly, the invention is not to be limited by what has been particularly shown and described, except as indicated by the appended claims or those ultimately provided in a utility application claiming priority to this provisional application.

CLAIMS:

1. A tomosynthesis method for creating a three-dimensional reconstruction of a target element volume comprising:

acquiring radiation absorbance images of the target element volume through a limited plurality of angles;

dividing the target element volume into a plurality of volume segments;

applying a reconstruction algorithm to each segment to generate a three-dimensional reconstruction of each volume segment; and

merging the three-dimensional reconstruction of each volume segment to create a three-dimensional reconstruction of the target volume.

2. The method of claim 1, wherein the volume segments overlap.

3. The method of claim 2, wherein the volume segments overlap by an amount sufficient to result in a three-dimensional reconstruction of the target volume that does not differ substantially in quality from an unsegmented reconstruction.

4. The method of claim 3, wherein the value of a majority of the pixels in the three-dimensional reconstruction of the target volume differ by less than about 1% from pixels in the unsegmented reconstruction.

5. The method of claim 2, wherein the volume segments overlap by between about 0 and 50 per cent.

6. The method of claim 1, wherein the volume segments comprise volume segments having a complex shape that is dependent upon the acquisition geometry and makes the reconstruction of each segment independent of the reconstruction of any other segment.

7. The method of claim 1, wherein the volume segments comprise volume segments having a slanted rectangular shape so that the reconstruction of each segment is independent of the reconstruction of any other segment.

8. The method of claim 1, wherein the images are obtained using an image acquisition element having:

a radiation source positionable at a plurality of angles in a first plane with respect to the target element; and

a radiation detector positioned so as to detect radiation emitted by the radiation source passing through the target element and determine a plurality of attenuation value for radiation passing through the target element to establish a radiation absorbance projection image of the target element for a particular radiation source angle.

9. The method of claim 8, wherein the volume segments comprise volume segments having a slanted rectangular shape and having a base that corresponds to a plurality of consecutive detector pixel rows that are parallel to and spaced apart from the first plane, each slanted rectangular volume segment extending from its base in a direction toward the radiation source.

10. The method of claim 8, wherein the method is deployed in mammography.

11. The method of claim 10, wherein the first plane is substantially parallel to a patient's chest wall.

12. The method of claim 11, wherein the volume segments comprise volume segments having a slanted rectangular shape and having a base that corresponds to a plurality of consecutive detector pixel rows that are parallel to and space apart from the patient's chest wall, each slanted rectangular volume segment extending from its base in a direction toward the radiation source.

13. A system for three-dimensional tomosynthesis imaging of a target element comprising:

an image acquisition element for obtaining a plurality of images of the target element from a plurality of angles having:

a radiation source positionable at a plurality of angles in a first plane with respect to the target element; and

a radiation detector positioned so as to detect radiation emitted by the radiation source passing through the target element and determine a plurality of attenuation value for radiation passing through the target element to establish a radiation absorbance projection image of the target element for a particular radiation source angle; and

a processor configured to apply a reconstruction algorithm to the radiation absorbance projection images of the target element obtained from a plurality of radiation source angles to generate a three-dimensional reconstruction of the target element wherein the processor is further configured to divide the target volume into a plurality of image reconstruction volume segments for separate image reconstruction of the volume segments and mergence of the reconstructed volume segments into a three-dimensional reconstruction of the target element.

14. A computer program for three-dimensional tomosynthesis imaging of a target element volume from a plurality of radiation absorbance projection images obtained at a different angles from an image acquisition element having a radiation source positionable at a plurality of angles within a first plane with respect to the target element and a radiation detector positioned so as to detect radiation emitted by the radiation source passing through the target element volume and determine a plurality of attenuation value for radiation passing through the target element to establish a radiation absorbance projection image of the target element volume for a particular radiation source angle, the computer program code being embodied in a computer readable medium and comprising computer program code for:

dividing the target element volume into a plurality of volume segments;

applying a reconstruction algorithm to each segment to generate a three-dimensional reconstruction of each volume segment; and

merging the three-dimensional reconstruction of each volume segment to create a three-dimensional reconstruction of the target volume.

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